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(56) Documents Cited
US 5109198 A US 4837515 A US 4638253 A

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(54) **High-Q and high homogeneity NMR birdcage resonator**

(57) A high-Q birdcage resonator for use as a receiver or transmitter in NMR or MRI has a reduced number of chip capacitors 2 disposed in each end-ring. Because there is a loss associated with each capacitor, the reduction in the number of capacitors raises resonator Q-factor. There is a plurality of parallel axial conducting strips 3 between each pair of circumferentially adjacent capacitors 3 on the end-rings. The width of the strips and their distribution is arranged to improve the RF field homogeneity in the resonator. The shielding rings 4 reduce electric field coupling to the sample. The RF port is preferably coupled between two conducting strips.

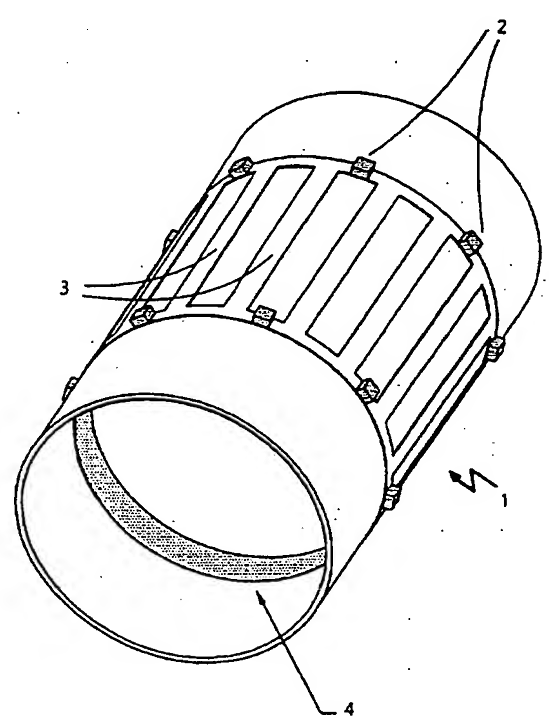

Figure 6

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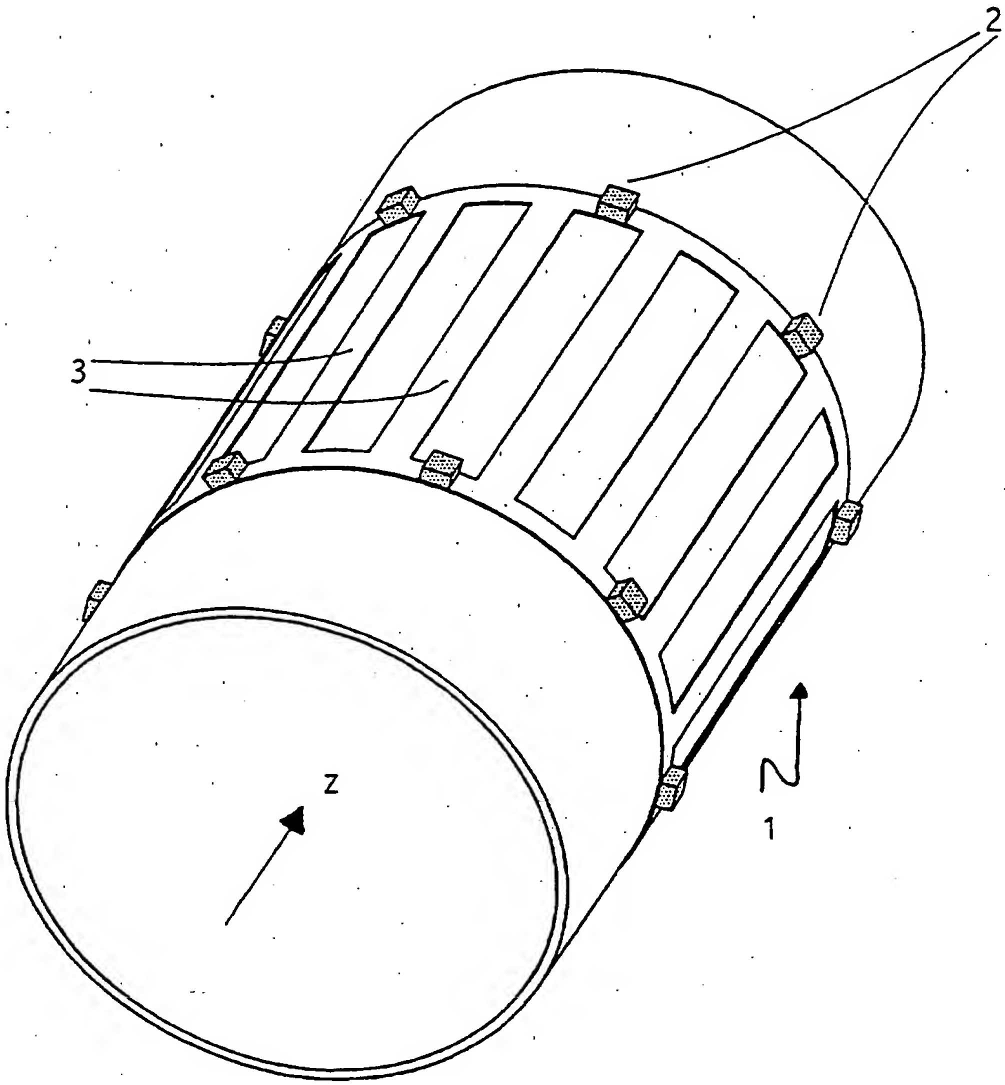


Figure 1

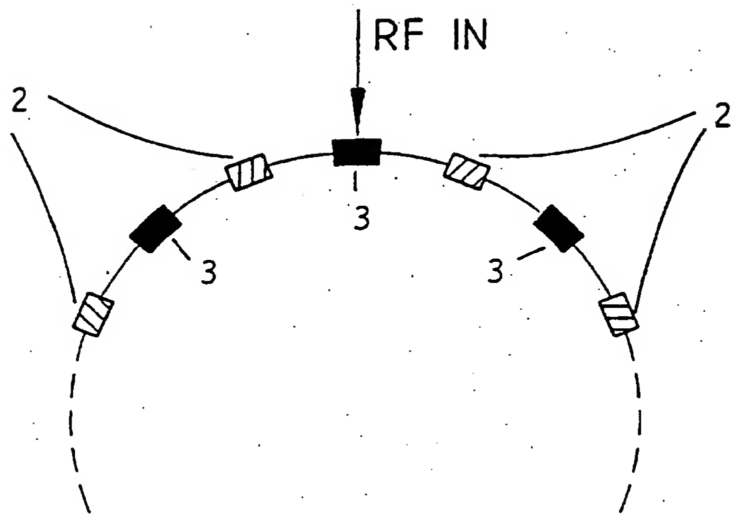


Fig. 2a

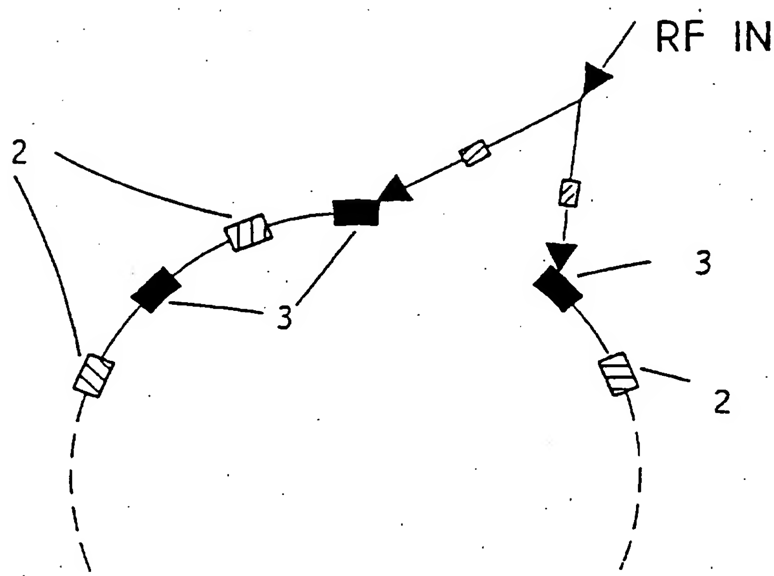


Fig. 2b

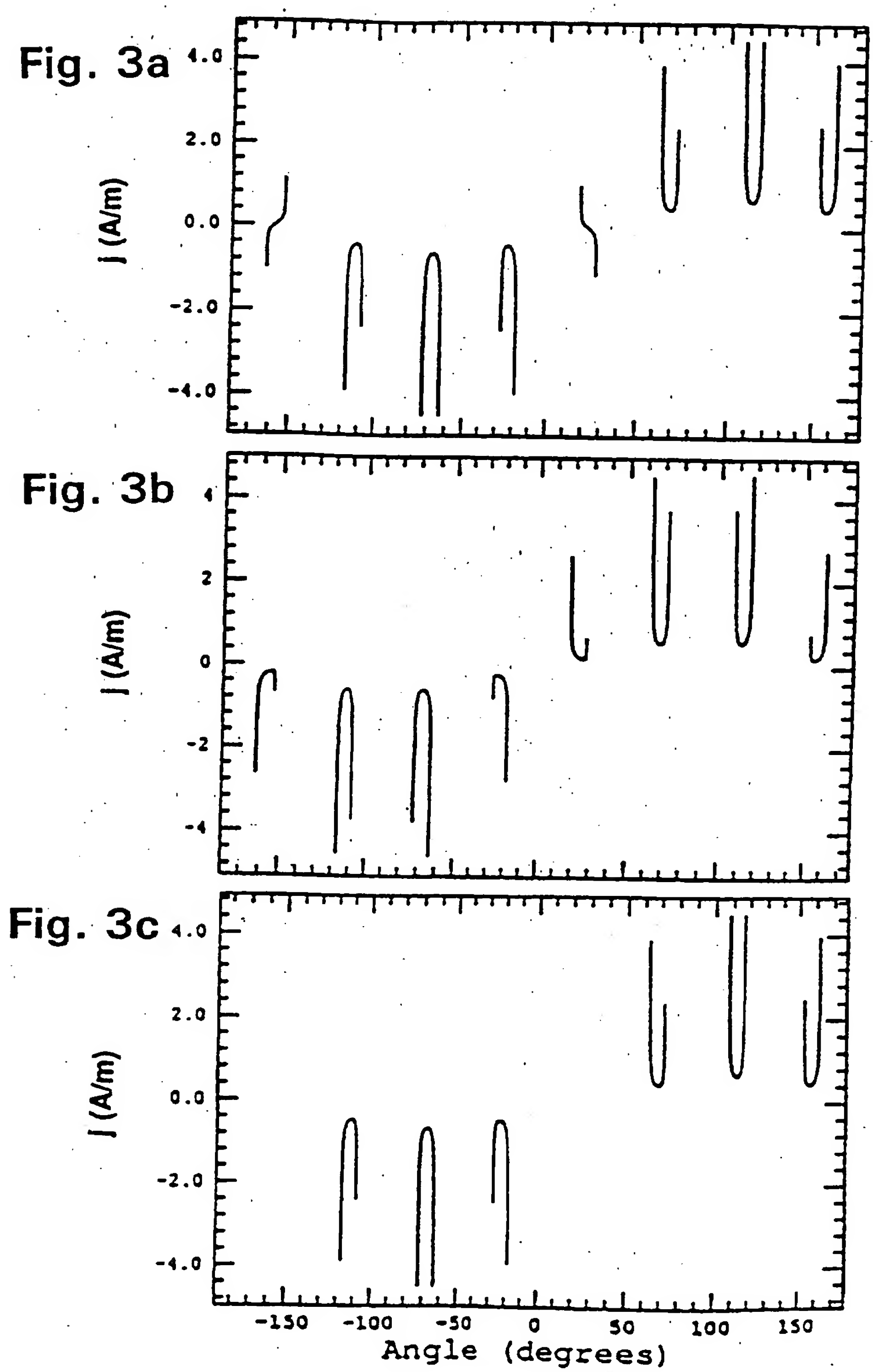


Fig. 3a
Fig. 3b
Fig. 3c
J (A/m)
J (A/m)
J (A/m)
4.0
2.0
0.0
-2.0
-4.0
4
2
0
-2
-4
-150
-100
-50
0
50
100
150
Angle (degrees)

Fig. 4a

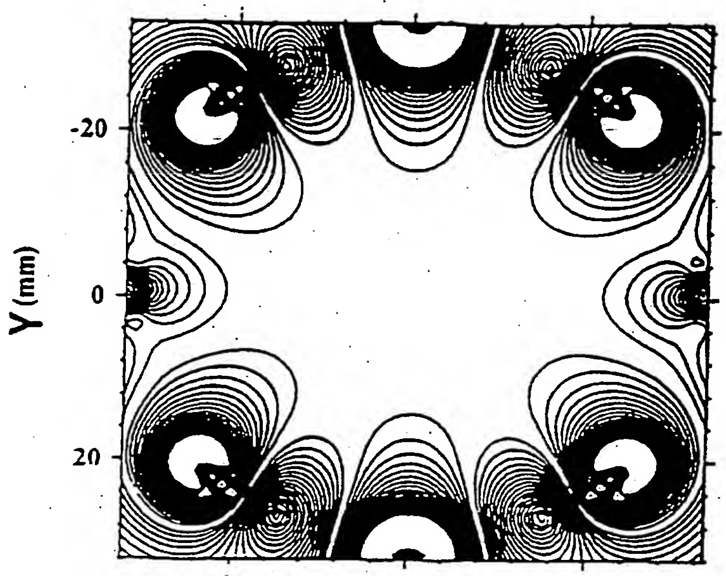


Fig. 4b

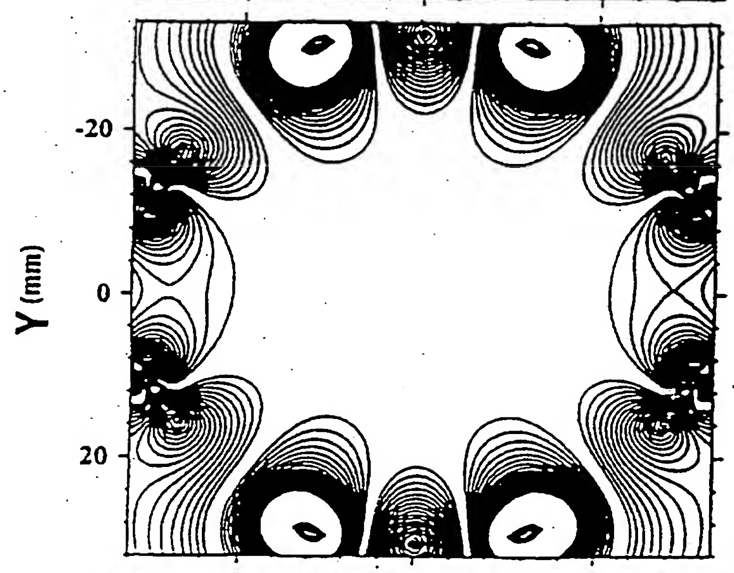


Fig. 4c

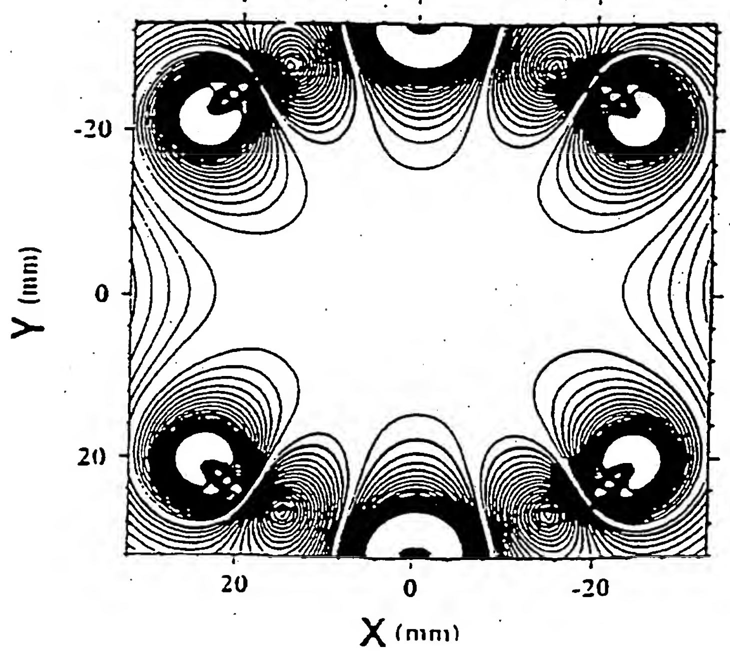


Fig. 5a

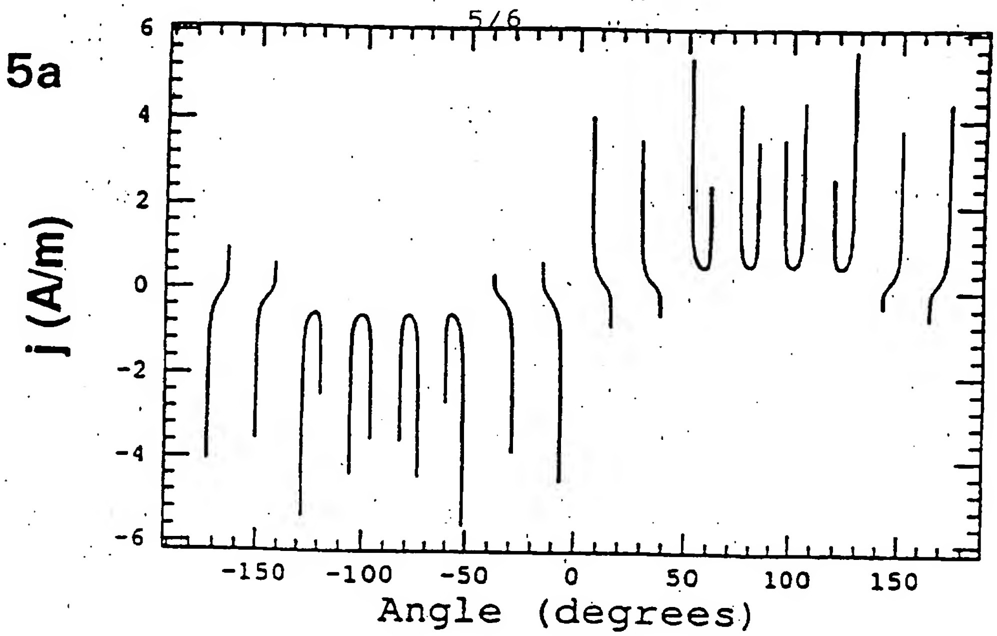
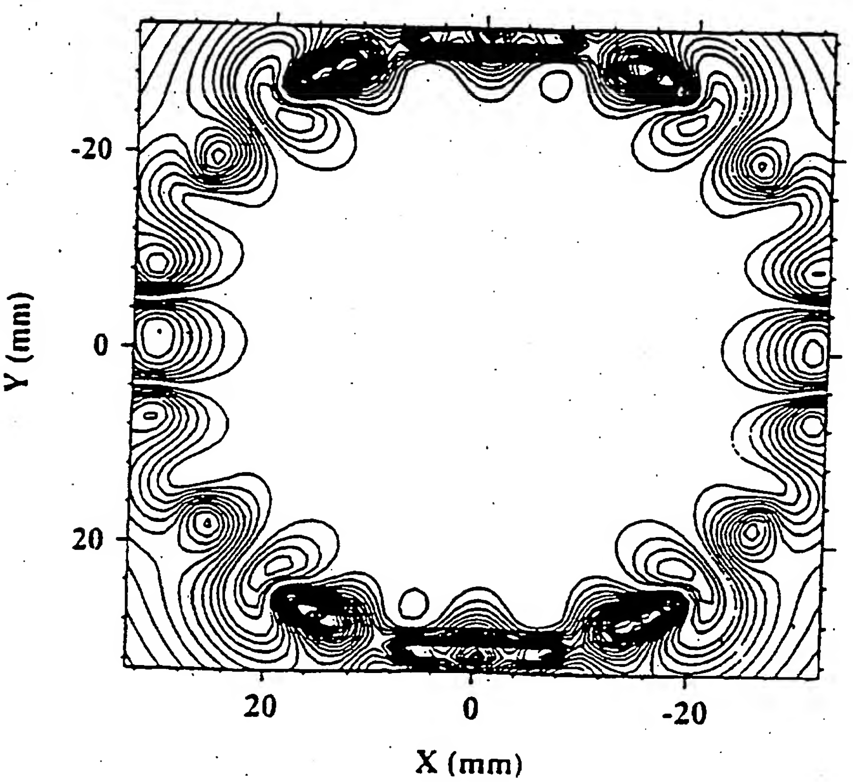

j (A/m)
6
4
2
0
-2
-4
-6
-150
-100
-50
0
50
100
150
Angle (degrees)

Fig. 5b


Y (mm)
-20
0
20
20
0
-20
X (mm)

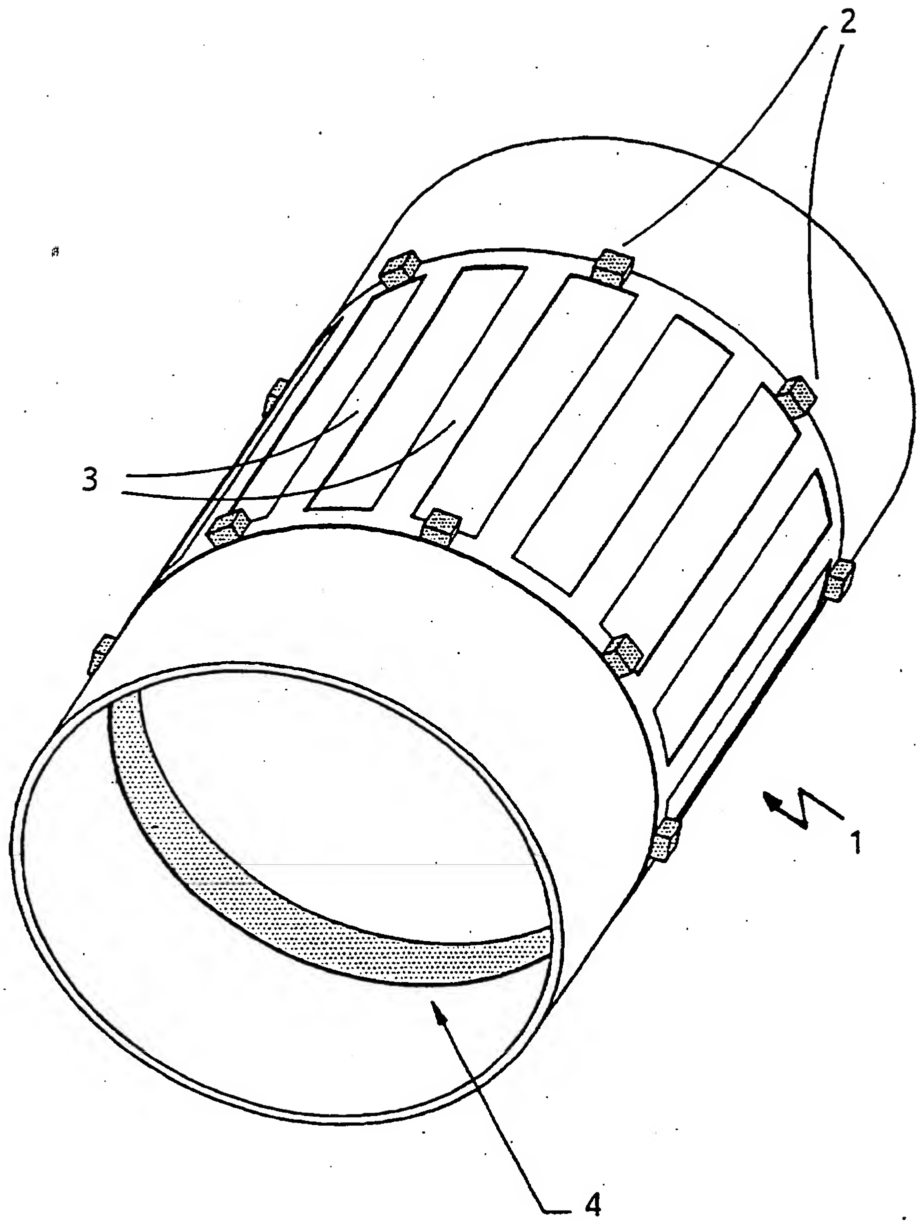


Figure 6

AN RF RESONATOR FOR NMR

BACKGROUND OF THE INVENTION

This invention relates to an RF excitation and/or receiver for use in an NMR apparatus, the probe comprising a generally tubular member acting as an RF resonator and having a plurality of circumferentially spaced axial conductors extending between a pair of spaced ring-like conductors, and a plurality of capacitive elements spaced along and interrupting the ring-like conductors. In particular the invention concerns a device for irradiating a sample with radio frequency (RF) energy and receiving NMR signals from it. The device may be used as either a transmitter or receiver, or both.

A device of this kind is, for example, known in the art from EP 0 177 855 B1.

The principal governing relationship in NMR is the Larmor equation:

$$\omega = \gamma B_0$$

where ω is the Larmor precessional frequency, γ is the nuclei specific gyromagnetic ratio and B_0 is the applied magnetic field. This equation applies to the situation where an ensemble of nuclei possessing nuclear spin are subjected to a strong magnetic field. A number of possible energy levels are developed by the interaction of the nuclear spins (which possess magnetic moments) and the applied field. In order to induce transitions between these energy levels, RF energy (B_1 field) is applied to the ensemble at the Larmor precessional frequency, with a B_1 direction orthogonal to

the direction of the applied field.

After the RF excitation is removed or ceases, the spin ensemble tends to return to its original state and in doing so emits energy. This is the received NMR signal. This signal can be detected by the same device (termed an RF probe) that was used to transmit the RF excitation, or by a separate probe. The (or each) probe normally comprises a coil or coil-like structure. In either case the probe(s) is/are tuned to, or near to the Larmor frequency.

It is most important in NMR and magnetic resonance imaging (MRI) experiments to maximize the signal-to-noise ratio (SNR) of the experiment, and to irradiate all parts of the sample with the same strength RF field. Similarly, it is important that the NMR signal from all parts in the sample be received by the RF probe with the correct weighting. Perhaps the two most important characteristics of an RF probe are the provision of a homogeneous B_1 field in the volume of the probe coil, and the possession of a high quality factor (Q). By reciprocity, if a coil provides homogeneous excitation it will also receive NMR signals in a homogeneous fashion. In this specification it will be assumed that discussions of excitation distributions of coils apply with equal relevance to their use as NMR receivers.

The Q of a coil is defined as 2π times the ratio of the time-averaged stored energy in the cavity to the energy loss per cycle. The Q of a coil has a profound effect on the SNR of the NMR experience (SNR α $(Q)^{1/2}$).

Prior art probes have been designed to provide a homogeneous

B_1 field without regard to Q, or a high Q coil without regard to RF field homogeneity, since optimization of one is usually at the expense of degradation of the other.

It is an object of this invention to provide an RF coil that provides both a substantially homogeneous RF field and a high quality factor.

It is a preferred object of this invention to provide an RF coil which optimizes both the homogeneity of the RF field and the Q of the probe for a particular situation.

SUMMARY OF THE INVENTION

In accordance with the invention the number of capacitive elements on each of the ring-like conductors is smaller than the number of axial conductors.

In a broad form, the invention provides an improved RF excitation and/or receiver probe for use in an NMR apparatus, the probe comprising a generally tubular member having a plurality of circumferentially spaced axial conductors extending between a pair of spaced ring-like conductors, and a plurality of capacitative elements spaced along the ring-like conductors.

The tubular member serves as the RF coil or "resonator".

In a first aspect of the invention, the number of capacitative elements in the coil is limited to avoid significant fall-off in the quality factor of the coil. The maximum number of capacitors to be used is determined from

the size of the coil and its frequency of operation.

In another aspect of the invention, the coil is characterized in that it has a plurality of axial conductors between each adjacent pair of capacitors. It has been found that using a parallel array of rungs between each adjacent pair of capacitors, and varying the width of the rungs, results in better approximation of the desired current density.

In yet another aspect of the invention, RF energy is fed to the probe between axial conductors thereof. This inter-rung feeding arrangement improves the RF field homogeneity within the coil.

Preferably, guard rings are placed inside the ring-like conductors of the coil to limit the RF window of the coil and minimize degradation of the quality factor of the coil by reducing the losses resulting from electric field linking to the sample within the coil.

The RF probe can advantageously comprise a plurality of axial conductors between each pair of adjacent capacitive elements on the ring-like conductors.

The RF probe can advantageously have exactly two axial conductors between each pair of adjacent capacitive elements on the ring-like conductors.

The RF probe can advantageously comprise 16 axial conductors.

The RF probe can advantageously have less than 9 capacitive

elements on each of the ring-like conductors.

The RF probe can advantageously have an operating frequency equal to or greater than 200 MHz, in particular 300 MHz.

The RF probe can advantageously have axial conductors which are conducting strips of finite widths.

The RF probe can advantageously have an inner diameter of the tubular member of less than 100 mm, in particular 64 mm.

The RF probe can advantageously have axial conductors of different widths.

The RF probe can advantageously have axial conductors distributed azimuthally in such a way that a sinusoidal current distribution is optimally approximated.

The RF probe can advantageously have the RF energy fed to the probe or extracted from the probe between axial conductors thereof.

The RF probe can advantageously have conductive guard rings placed internal to the capacitors to limit the RF window of the probe and to lower dielectric losses to a sample inside the tubular member.

In order that the invention may be more fully understood and put into practice, a preferred embodiment thereof will now be described with reference to the accompanying drawings.

BRIEF DESCRIPTION OF THE DRAWING

In the drawings,

Fig. 1 is a perspective view of a coil resonator according to one embodiment of the invention;

Fig. 2a illustrates feeding of a resonator coil according to prior art;

Fig. 2b illustrates inter-rung feeding of a resonator coil according to the invention;

Fig. 3a illustrates the current density distribution from a standard rung fed design;

Fig. 3b illustrates the current density distribution from an inter-rung fed device with no rings carrying zero integrated current;

Fig. 3c illustrates the current density distribution showing the effect of removing rungs completely from the structure;

Fig. 4a illustrates the transverse RF field distribution for the coil of Fig. 3a;

Fig. 4b illustrates the transverse RF field distribution for the coil of Fig. 3b;

Fig. 4c illustrates the transverse RF field distribution for the coil of Fig. 3c;

Fig. 5a shows the current density distribution for the

inter-rung hybrid coil design of Fig. 1; and

Fig. 5b shows a transverse field deviation for the inter-rung hybrid coil design of Fig. 1.

Fig. 6 is a perspective view of a resonator coil having a guard ring according to a further embodiment of the invention.

An RF coil or resonator typically consists of a number of inductive structures distributed around a tube of circular cross-section. In order to generate a homogeneous transverse RF field, it is necessary to establish a current distribution of longitudinal current (the longitudinal direction being defined by the axis of the resonator) such that the current varies as sin θ, where θ is the azimuthal angle. This is discussed in prior art US patent no. 4,694,255 and EP 0 177 855 B1 the disclosure of which are herein incorporated by reference.

Typically, this current distribution is established by creating a standing wave around the periphery of the coil by separating each rung by an appropriate capacitor, and thereby also creating a tuned structure, or by having capacitors in each longitudinal element which would then be connected at the top. Such arrangements are commonly known as "birdcage" resonators (details of which are given in Journal of Magnetic Resonance, 63,622 (1985)).

As the number of capacitors and inductive elements is increased around the structure, the desired sin θ current distribution is better approximated. The conventional design process is to include as many of these structures into the

coil as possible and still tune the coil to the desired frequency. However, a large number of capacitors in the structure results in large energy losses due to both reactive and non-reactive losses in the capacitors. This is partly due to the capacitors experiencing a very large electric field resulting in significant dielectric losses. Thus, although increasing the number of capacitors improves homogeneity, the quality factor decreases significantly.

In this invention, unlike conventional design, the number of capacitors is limited to maintain quality factor yet still achieve a sufficient degree of homogeneity. To illustrate the effect on specific coil designs, we will detail designs of a coil with an internal diameter of 64 mm, although the designs discussed here may be equally applied to coils of any dimension.

Figure 1 illustrates a preferred embodiment of a resonant structure suitable for use in an NMR probe head. The number of capacitors is restricted at a particular operating frequency in order to reduce losses in the resonator structure, and in between each capacitor is an array of conductors, distributed so as to best mimic the desired current density.

The maximum number of capacitors to be used depends on the size of the coil and its frequency of operation. For coils with an internal diameter of 64 mm, a significant fall-off in Q occurs when the number of capacitors exceeds eight. The quality and indeed the quality factor of a capacitor is a frequency dependent value; manufacturers generally provide performance specification curves that correlate capacitor value and frequency of operation with the Q and Effective

Series Resistance (ESR) of the capacitors. The ESR is a value that includes all losses in the capacitor. To minimize this value, high performance capacitors are used and the number of capacitors (and concomitant losses) is reduced.

The rung distribution is optimized by introducing a parallel array of rungs between each pair of capacitors and varying the width of the rungs. The distribution of the conductors is such that they more closely generate the desired current density.

As shown in Figure 1, the coil 1 comprises inductive rungs 3 running parallel to the Z axis between adjacent pairs of capacitors 2. Although only two rungs 3 connected in parallel are shown here for clarity, many could be used. When a large number of inter-capacitor rungs are used, their azimuthal spacing is chosen such that they best approximate that segment of the standing wave quantized by the capacitor.

Further to the novel rung arrangement, an inter-rung feeding scheme is used to provide additional improvement to the RF field homogeneity. The scheme is shown schematically in Figure 2b, in contrast to the conventional method of rung feedings shown schematically in Figure 2a. These figures are cross-sections (in the Y plane) of Figure 1. This is a capacitive coupling feeding arrangement and prior to the feed point shown (labelled RF in) are the normal variable tuning and matching capacitors being fed from an RF amplifier.

To illustrate the effect of inter-rung feeding, the resultant current densities from a conventional eight rung

resonator and that from an inter-rung fed design are shown in figures 3a, 3b, and 3c. In these examples, the rung width was 5 mm and the frequency of operation was 300 MHz.

It will be assumed for simplicity in the present analysis that field and current variations are purely sinusoidal in the z-direction so that the Transverse Electric and Magnetic (TEM) approximation can be invoked. It follows that the electric field **E**, the magnetic induction field **B** and the current density **j** per unit cross-sectional width of the conducting strip can be expressed in the (approximate) forms

$$\begin{aligned} \mathbf{E}(x,y,z,t) &= \mathbf{E}_T(x,y)\exp(i\omega[\sqrt{\mu\varepsilon}z - t]) \\ \mathbf{B}(x,y,z,t) &= \mathbf{B}_T(x,y)\exp(i\omega[\sqrt{\mu\varepsilon}z - t]) \\ \mathbf{j}(x,y,z,t) &= \mathbf{j}_T(x,y)\exp(i\omega[\sqrt{\mu\varepsilon}z - t]), \end{aligned} \tag{1}$$

where the magnetic permeability and electric permittivity of the air surrounding the conducting strips are μ and ε, respectively, the angular frequency of the signal is ω, and t denotes time. In this TEM approximation, the transverse parts $\mathbf{E}_T$ and $\mathbf{B}_T$ of the electric and magnetic fields have no axial component, so that

$$\mathbf{E}_T \cdot \mathbf{k} = 0 \qquad \text{and} \qquad \mathbf{B}_T \cdot \mathbf{k} = 0,$$

and **k** denotes the unit vector pointing in the z-direction.

A consequence of the assumption (1) of TEM-mode solutions is that the full system of governing equations (Maxwell's equations) possesses solutions in which there is a simple relationship between the magnetic and electric fields, given by

$$\mathbf{B}_T = -\sqrt{\mu\varepsilon}\,(\mathbf{k} \times \mathbf{E}_T). \tag{2}$$

Furthermore, it follows from equations (1) and (2) that Faraday's law reduces to

$$\nabla_2 \times \mathbf{E}_T = 0,$$

from which a scalar potential Φ can be defined immediately for the electric field, according to the relation $\mathbf{E}_T = -\nabla_2\Phi$. Here, $\nabla_2 = (\partial/\partial x, \partial/\partial y)$ is the gradient operator in the transverse plane.

For the purposes of computing the electric and magnetic fields within the probe and the current densities within the longitudinal rungs, it will be assumed that the copper rungs in the resonator are perfect conductors. The boundary condition to be imposed is therefore that Φ must be constant along the surface of each conductor.

Once the scalar potential Φ has been determined, the transverse part $\mathbf{j}_T$ of the current density at the surface of the conductor may be determined according to

$$\mathbf{j}_T = \sqrt{\mu/\varepsilon}\,(\mathbf{n}\cdot\nabla_2\Phi)\mathbf{k}, \tag{3}$$

where $\mathbf{n}$ represents the normal to the conducting surface.

Since the interior of the MRI probe does not possess sources of charge, Maxwell's equations also reveal that the transverse part of the electric field is solenoidal, so that $\nabla_2\cdot\mathbf{E}_T = 0$, under the TEM approximation (1) and the relationship (2). It follows that the scalar potential Φ

satisfies Laplace's equation $\nabla_2{}^2\Phi = 0$.

A numerical algorithm based on the Inverse Finite Hilbert Transform was used to obtain the final current density after applying the appropriate boundary conditions. Figure 3a shows the current density resulting from a standard rung fed design. In such a design, two of the rungs carry zero integrated current. Figure 3c shows the effect of removing such rungs from the structure entirely (which is done for mode-stabilization). Figure 3b shows the resultant current density from an inter-rung fed device in which no rungs carry zero integrated current.

The transverse RF fields:

$$B_T = \sqrt{B_x{}^2+B_y{}^2}$$

were then calculated from the current densities for each coil arrangement. The transverse fields corresponding to the current densities of Figs. 3a, 3b, and 3c are shown in Figs. 4a, 4b, and 4c. The fields are shown as contour plots of the deviation from a perfectly homogeneous field, with each contour representing a 5% level away from the normalized central region. The area contained by the first contour then represents a field with a deviation from homogeneity of less than 5%; similarly the area contained by the second contour from the centre represents 10% or better homogeneity, etc. The fields were calculated in 1 mm intervals across the structure at 300 MHz.

The drawings show significantly improved homogeneity in the central region of the coil that used inter-rung feeding (Figure 4b).

Figs. 5a and 5b depict the current density and field plots, shown as 5% contours, for an inter-rung hybrid coil design having a parallel array of rungs between each adjacent set of capacitors as shown in Figure 1. These results demonstrate the further improvement in field homogeneity while maintaining Q. A coil of these dimensions was built and its Q was measured to be 190. Measurements were made on a HP8711A network analyzer at 3 dB points. An equivalent conventional coil of 16 rungs and 16 capacitors has a measured Q of 85. The described design provides improved performance over conventional designs.

In a further embodiment, guard rings 4 (rings of copper or other RF opaque conductors) are placed internal to the capacitors as shown in Figure 6 to limit the RF window of the coil and to lower the losses to the sample by electric field linking to the sample, the electric field being very high in the regions surrounding the capacitors. These rings also have the effect of preventing broad signals from being acquired from long samples extending outside the "well-shimmed" region of the magnet of the NMR apparatus.

The foregoing describes only some embodiments of the invention, and modifications which are obvious to those skilled in the art may be made thereto without departing from the scope hereof.

Claims

1. RF excitation and/or receiver for use in an NMR apparatus, the probe comprising a generally tubular member acting as an RF resonator and having a plurality of circumferentially spaced axial conductors extending between a pair of spaced ring-like conductors, and a plurality of capacitive elements spaced along and interrupting the ring-like conductors,

   characterized in that

   the number of capacitive elements on each of the ring-like conductors is smaller than the number of axial conductors.

2. RF probe according to claim 1, characterized in that it comprises a plurality of axial conductors between each pair of adjacent capacitive elements on the ring-like conductors.

3. RF probe according to one of the preceding claims, characterized in that there are exactly two axial conductors between each pair of adjacent capacitive elements on the ring-like conductors.

4. RF probe according to one of the preceding claims, characterized in that it comprises 16 axial conductors.

5. RF probe according to one of the preceding claims, characterized in that on each of the ring-like

conductors there are less than 9 capacitive elements.

6. RF probe according to one of the preceding claims, characterized in that its operating frequency is equal to or greater than 200 MHz, in particular 300 MHz.

7. RF probe according to one of the preceding claims, characterized in that the axial conductors are conducting strips of finite widths.

8. RF probe according to one of the preceding claims, characterized in that the inner diameter of the tubular member is less than 100 mm, in particular 64 mm.

9. RF probe according to one of the preceding claims, characterized in that the axial conductors have different widths.

10. RF probe according to one of the preceding claims, characterized in that the axial conductors are distributed azimuthally in such a way that a sinusoidal current distribution is optimally approximated.

11. RF probe according to one of the preceding claims, characterized in that the RF energy is fed to the probe or extracted from the probe between axial conductors thereof.

12. RF probe according to one of the preceding claims, characterized in that conductive guard rings are placed internal to the capacitors to limit the RF window of the probe and to lower dielectric losses to a sample inside the tubular member.

13. RF probe substantially as hereinbefore described with reference to and as illustrated by Figures 1 to 6 of the accompanying drawings.

The Patent Office

17

Application No: GB 9603648.8
Claims searched: 1-13

Examiner: K. Sylvan
Date of search: 14 May 1996

Patents Act 1977
Search Report under Section 17

Databases searched:

UK Patent Office collections, including GB, EP, WO & US patent specifications, in:

UK Cl (Ed.O): G1N (NG34)

Int Cl (Ed.6): G01R (33/34)

Other:

Documents considered to be relevant:

Category	Identity of document and relevant passage		Relevant to claims
X	US5109198	Hitachi. See figure 14.	1,2,4,5,12
X	US4837515	Mitsubishi. See the figures and column 3 line 62 to column 4 line 6.	1,2,4,5,7, 12
X	US4638253	GE. See figures 2 and 3.	1,5,10

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